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Sagittal plane body kinematics and kinetics during single-leg landing from increasing vertical heights and horizontal distances: Implications for risk of non-contact ACL injury

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ABSTRACT

Purpose: This study identified kinematic and knee energetic variables that reduce the risk of non-contact anterior cruciate ligament (ACL) injury during single-leg landings from increasing vertical heights and horizontal distances.
Methods: Nine subjects performed single-leg landings from takeoff platforms with vertical heights of 20, 40, and 60 cm onto a force plate. Subjects also performed single-leg landings from a 40 cm high takeoff platform placed at horizontal distances of 30, 50 and 70 cm from a force plate. Kinematic and kinetic data were measured.
Results: Vertical height had a significant and positive effect on peak vertical ground reaction force (VGRF) ($p<0.001$), peak posterior ground reaction force (PGRF) ($p=0.004$), knee flexion angle ($p=0.0043$), trunk flexion angle ($p=0.03$), knee power ($p<0.001$) and knee work ($p<0.001$). There was also a significant and positive effect of horizontal distance on peak PGRF ($p<0.001$), ankle plantar flexion angle ($p=0.008$), hip flexion angle ($p=0.007$), and trunk flexion angle ($p=0.001$). At increasing vertical height, peak VGRF was significantly correlated to ankle plantar flexion and knee flexion angles ($r=-0.77$, $p=0.02$ and $r=-0.78$, $p=0.01$, respectively). At increasing horizontal distance, peak PGRF was significantly correlated to ankle plantar flexion angle, knee power and knee work ($r=-0.85$, $p=0.003$; $r=0.67$, $p=0.04$; and $r=0.73$, $p=0.02$, respectively).
Clinical Relevance: A better understanding of the risk factors to non-contact ACL injury during single-leg landings from increasing vertical heights and horizontal distances can aid in the design of injury prevention regimen.

1. Introduction

In a jump landing event, the landing phase is more stressful to the ACL than the takeoff phase [1]. Single-leg landing is a common athletic maneuver performed from varying vertical heights and horizontal distances during sports such as basketball, volleyball, soccer, and badminton [2]. Most non-contact ACL injuries occur during sporting activities involving single-leg landings [3]. Vertical height and horizontal distance may have different effects on landing kinematics and kinetics, thus warranting different injury prevention and training strategies. Particularly important is the ability of the body's kinematics to attenuate the GRFs during landings from a single-leg. This knowledge can provide important insights into improving our understanding of the key factors resulting in increased risk of non-contact ACL injuries during single-leg landings. As well, out-of-plane motion such as landing on a single leg with the knee abducted has been shown to result in valgus collapse, subsequently increasing the risk of ACL injury [4–7]. However, it is not clear if valgus collapse causes ACL injury or occurs as a result of the ACL being injured [4].

An *in vivo* study conducted by Cerulli et al. [8] demonstrated that during a single-leg hopping task, the peak ACL strain occurred at the same instant as peak VGRF, suggesting that peak VGRF may be a likely predictor for establishing the risk of non-contact ACL injuries. Cerulli et al. [8] obtained an average peak ACL strain of $5.47 \pm 0.28\%$, which may be too low to cause ACL injury during single-leg landings given two studies [9,10] estimated that the ACL is able to tolerate strains of up to 20% before failure. Nonetheless, the *in vivo* strain to failure of the ACL during single-leg landings is unknown, and therefore relating strain values to ACL injury to date is not yet possible. However, other studies have also determined that landing with a high impact force may be a risk factor to ACL injury [1,11–15]. In addition, a recent study by Boden et al. [16] proposed that a lack of dissipation of GRFs at landing may be a factor in ACL injury. Given this, the current study uses peak VGRF to predict risk of non-contact ACL injuries. The VGRF is defined as the reaction to the force the body exerts on the ground in the vertical direction. In addition, many studies have also reported that an increase in PGRF requires an increase in knee extensor moment for balance, and this increase in knee extensor moment is a major contributor to higher peak proximal tibia anterior shear force

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that likely increases ACL loading [15,17–19]. Given this, the current study also uses peak PGRF to predict the risk of non-contact ACL injuries. The PGRF is defined as the horizontal reaction force the body exerts on the ground in the backward direction from the landing direction.

The literature has shown that single-leg landing results in increased risk of non-contact ACL injuries compared to double-leg landing [20,21]. Pappas et al. [20] determined that single-leg landings resulted in greater risk of ACL injury compared to double-leg landings given the former led to significantly greater knee valgus, as well as, significantly lower knee flexion angles. The study by Yeow et al. [21] showed that single-leg landings produced significantly greater peak VGRF, as well as, significantly lower knee flexion angles and knee joint power compared to double-leg landings, implicating greater risk of ACL injury. There are many single-leg landing studies investigating non-contact ACL injuries [20–34]. Of these studies, many investigated single-leg landings from only one vertical height [20,22–30,32–34], and to the authors' best knowledge, no study has yet investigated the effect of increasing horizontal distance on single-leg landing biomechanics and further related these findings to risk of non-contact ACL injury. In addition, many single-leg landing studies investigated vertical heights lower than 30 cm, which as argued by Zhang et al. [35] is too low, and the subject's landing strategies may be completely different at greater heights. Altogether, existing single-leg landing studies have illustrated that landing technique, gender effects or fatigue may predispose athletes to ACL injury. However, it is still not clear how the body will respond in terms of kinematics and knee energetics to the effect of increasing vertical height and horizontal distance during single-leg landings. As well, in many single-leg landing studies, ankle, hip and trunk kinematics were not considered as direct contributors to risk of ACL injury. Moreover, simultaneously investigating ankle, knee, hip and trunk kinematics may be important for predicting risk of non-contact ACL injury given it has been shown that the vast majority of studies do not account for the effects of whole body movement on knee loads [36]. Pandy and Sasaki [37] and Griffin et al. [38] argued that the knee is just one part of the kinetic chain, and that the trunk, hip, and ankle may all contribute to risk of non-contact ACL injury. Though many single-leg landing studies predominately investigated knee kinematics and kinetics, further investigations are required to determine the coupling between trunk, hip, knee and ankle kinematics, and the link between these kinematic variables, knee energetics, ground reaction forces, and subsequently, risk of non-contact ACL injury.

The primary objective of this study was to identify kinematic and knee energetic variables that reduces the risk of non-contact ACL injury during single-leg landings, by examining the relationships among increasing vertical height and increasing horizontal distance and two non-contact ACL injury risk predictor variables, namely, peak VGRF and peak PGRF. To undertake this, we first investigated the effect of increasing vertical height and horizontal distance on peak VGRF, peak PGRF, ankle, knee, hip and trunk flexion angles, as well as knee energetics during single-leg landing. This study then correlated the variables significantly impacted by the main effect of vertical height or horizontal distance to two ACL injury risk predictor variables so as to assess the risk of non-contact ACL injury during single-leg landing. We hypothesized that vertical height would have a significant effect on knee flexion angle, as well as, knee power and knee work. We further hypothesized that as vertical height increased, there would be a significant correlation between peak VGRF and knee flexion angle.

2. Methods

2.1. Participants

Nine male recreational athletes with mean (SD) age of 27.67 (2.56) years, heights of 1.75 (0.077) m, and masses of 78.12 (8.59) kg were recruited from the university population. None of the participants reported any musculoskeletal or ligamentous injuries to the lower extremity at the time of participation. Prior to data collection, each participant gave informed consent as stipulated by the university ethics review board. Subjects' ages and anthropometrics were recorded. The dominant leg was established as the leg used by the subject to kick a ball.

2.2. Procedure

All participants wore identical shoes (running shoe, model BY004, ASICS America Corporation, Irvine, CA) throughout data collection, so as to mitigate variability. Retro-reflective markers were placed on subjects' body using a customized version of the Vicon Plug-in Gait marker set via double-sided tape. The Vicon Plug-in Gait marker set was customized to include additional markers at the hip and medial aspects of the elbow, knee and ankle, as well as additional foot markers. In addition, different marker locations were also used at the proximal ends of the pelvis. A seven-camera motion capture system (Vicon MX, Oxford Metrics, UK) was used to collect the subject's kinematics at a sampling rate of 250 Hz. A force plate (Kistler type 9281B, Winterthur, Switzerland) measured GRF signals at a sampling rate of 1000 Hz. Videographic and force plate data were time synchronized. Before data collection, each subject was given enough time to warm-up and practice the single-leg landing tasks until comfortable. The warm-up and practice regimen was standardized to mitigate possible variability contributed by such tasks. The subjects were instructed to stand on various takeoff platforms, each having vertical heights of 20, 40, and 60 cm (termed h20, h40 and h60, respectively), which were placed 20 cm from the rear edge of the force plate. This test setup was used to investigate the effect of increasing vertical height and termed the vertical height test. A 40 cm high takeoff platform was placed at a horizontal distance of 30, 50 and 70 cm (termed d30, d50 and d70, respectively) from the rear edge of the force plate to test the effect of increasing horizontal distance, and was termed the horizontal distance test.

The subjects were instructed to stand on the takeoff platforms with both arms placed on their iliac crests and with legs spaced approximately shoulder width apart. From this position, the subjects were asked to stand on their dominant leg, jump forward, and land as naturally as possible with their dominant leg only onto the force plate. Each subject performed two trials at each vertical height and horizontal distance resulting in a total of 12 trials. The order of the heights and distances of landing were randomized to reduce learning effects.

2.3. Data reduction and analysis

All marker trajectories and analog data were imported into Visual3D (C-Motion Inc. Rockville, MD) biomechanical software for rigid body modeling and inverse dynamics analysis (IDA). Cardan angles for the hip, knee, and ankle were calculated in an x (flexion–extension), y (adduction–abduction), z (internal–external rotation) sequence. In Visual3D ankle dorsiflexion is defined as positive, knee flexion as negative, hip flexion as positive, and trunk flexion as positive. Ankle angle was defined as the angle between the leg segment and foot segment. The difference between ankle joint angle when subject was standing on the takeoff platform and when peak VGRF occurred during landing was determined as the ankle flexion angle. Knee flexion angle was defined as the angle between the thigh and leg segments while hip flexion angle was defined as the angle between the thigh and pelvis segments. Trunk flexion angle was calculated as the angle between the trunk segment and a vertical line in the laboratory coordinate system. All kinematic, kinetic and knee power values were determined at the instant of peak VGRF. Kinematic data were low-pass filtered using a second-order bidirectional Butterworth filter at 6 Hz and analog data were filtered at 25 Hz. Joint kinetic data were calculated using a complete Newtonian IDA.

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One trial was selected from the better of two trials for model building, data analysis, post-processing, and reporting. The better trial was determined as the one in which the participant did not remove his hands from the iliac crests during landing, did not land with both legs on the force plate, and did not lose a marker during impact with the ground. In situations where both trials did not meet these requirements, the subject repeated the task until a trial that met these requirements was obtained. Furthermore, in situations where both trials met these requirements, a trial was arbitrarily selected for post processing and data analysis. Knee joint energetics such as knee power and knee work are important indicators of energy dissipation during the landing phase. Knee power was determined by the product of knee moment and knee angular velocity over the landing phase. Knee work was then calculated as the integral of the negative portion of the knee power curve over the landing phase. The use of the negative portion of the power curve over the landing phase was based on studies [23,39] that showed the negative power curve represents the energy dissipation by the knee extensor moment. Knee power and knee work were normalized to body mass. The landing phase was defined as 0.8 s prior to peak VGRF and 0.6 s post peak VGRF, where the time of peak VGRF was used as the second event, and 0.8 s prior and 0.6 s after peak VGRF were used as the first and third event, respectively. Kinematic data were time normalized to 100% of the landing phase (between the first and third event). Ground reaction forces were normalized to each subject's body weight in newtons (N), while knee energetics were normalized by the subject's body mass in kilograms (kg).

2.4. Statistical analysis

Ground reaction forces, sagittal plane body kinematics, as well as knee power and knee work were averaged across all subjects at each trial. Multiple one-way repeated-measure ANOVAs were conducted to test the effect of vertical height on all dependent variables (vertical height test). The dependent variables were peak VGRF, peak PGRF, ankle, knee, hip and trunk flexion angles, as well as, knee power and knee work. Post-hoc analyses were conducted via pairwise comparisons using dependent sample t-tests. For the vertical height test, this would result in three pairwise t-tests: h20 vs. h40, h20 vs. h60, and h40 vs. h60. The p-value for each test was Bonferroni corrected and compared to an alpha level of 0.016 (0.05/3). This ensures that the probability of committing a Type I error (rejecting the null hypothesis when it is true) will not be greater than 0.05 for the entire set of post-hoc analyses. Follow-up testing using Pearson product moment correlations (PPMCs) were also calculated to determine the relationships among the two ACL injury risk predictor variables, ankle, knee, hip and trunk flexion angles, as well as, knee power and knee work. Multiple one-way repeated-measure ANOVAs were also conducted to test the effect of horizontal distance on all dependent variables (horizontal distance test). Pairwise comparison using dependent sample t-tests were conducted for the horizontal distance test: d30 vs. d50, d30 vs. d70 and d50 vs. d70. Follow-up testing using PPMCs were also calculated to determine the relationships among the two ACL injury risk predictor variables, ankle, knee, hip and trunk flexion angles, as well as, knee power and knee work. Effect size is presented as partial η^2 (eta squared). Partial η^2 ranges between 0 and 1. These values can be interpreted using the following parameters: $0.01 \leq$ partial $\eta^2 < 0.06$ indicates a small effect, $0.06 \leq$ partial $\eta^2 < 0.14$ indicates a medium effect, and partial $\eta^2 \geq 0.14$ indicates a large effect [40]. To interpret Pearson's correlation coefficients (r), values of $|r| < 0.3$ indicate a weak correlation, $0.3 \leq |r| < 0.7$ indicate a moderate correlation and $|r| \geq 0.7$ indicate a strong correlation. Unless otherwise stated, an alpha level of 0.05 was used throughout to identify statistical significance. Statistical analyses were conducted in SPSS (Version 11.5, Chicago, IL, USA).

3. Results

The findings from the separate ANOVAs conducted, revealed for the vertical height test that there was a significant effect of vertical height on peak VGRF ($F(2,16) = 42.41$, $p < 0.001$, partial $\eta^2 = 0.84$, observed power = 1.0). Our findings also revealed a significant effect of vertical height on peak PGRF ($F(2,16) = 8.09$, $p = 0.004$, partial $\eta^2 = 0.505$, observed power = 0.91). There was also a significant effect of vertical height on knee flexion angle ($F(2,16) = 3.86$, $p = 0.043$, partial $\eta^2 = 0.33$, observed power = 0.61), trunk flexion angle ($F(2,16) = 12.21$, $p = 0.001$, partial $\eta^2 = 0.60$, observed power = 0.99), knee power ($F(2,16) = 24.99$, $p = 0.0001$, partial $\eta^2 = 0.76$, observed power = 1.0), and knee work ($F(2,16) = 64.07$, $p = 0.0001$, partial $\eta^2 = 0.89$, observed power = 1.0). Partial η^2 represents an index of strength of association between an experimental factor and the dependent variable [41]. Larger values of partial η^2 indicate a greater amount of variation accounted for by the dependent variable, to a maximum value of 1. The overall means and standard deviations of the sagittal plane body kinematics, knee power, and knee work for the vertical height test among all subjects are provided in Table 1. Figure 1A and B shows the trends in peak VGRF and peak PGRF, respectively, during single-leg landings for all subjects as vertical height increased. From Fig. 1A and B, one can glean that vertical height had a significant and positive effect on both peak VGRF and peak PGRF, suggesting that as vertical height increased, both the peak VGRF and peak PGRF increased. The differences in marginal means at the three vertical heights are (1.26, 0.73) and (0.07, 0.17), which represent the difference in peak VGRF and peak PGRF respectively, at landing heights of 20 to 40 cm and 40 to 60 cm, respectively.

For the vertical height test, pairwise dependent sample t-tests indicated that peak VGRF was significantly lower at h20 compared to h40 ($t(8) = 5.61$, $p < 0.016$), and h60 ($t(8) = -7.94$, $p < 0.016$). The peak VGRF was also significantly lower at h40 compared to h60 ($t(8) = -4.20$, $p < 0.016$). Peak PGRF was determined to be significantly higher at h60 compared to h20 ($t(8) = -3.50$, $p < 0.016$), and h40 ($t(8) = -3.29$, $p = 0.011$). There was no significant difference in peak PGRF at h20 compared to h40 ($t(8) = -1.16$, $p > 0.016$). Knee flexion angle was significantly higher at h60 compared to h20 ($t(8) = 2.42$, $p = 0.015$). There was no significant difference in knee flexion angle at h20 compared to h40, as well as, h40 compared to h60. Trunk flexion angle was significantly higher at h60 compared to h20 ($t(8) = -3.14$, $p = 0.014$) and h40, ($t(8) = -4.45$, $p = 0.002$). There was no significant difference in trunk flexion angle at h20 and h40 ($t(8) = 1.82$, $p > 0.016$). Knee power was significantly lower at h20 compared to h40 ($t(8) = 4.26$, $p = 0.003$) and h60, ($t(8) = 7.14$, $p < 0.016$). There was no significant difference in knee power at h40 compared to h60. Results indicated that knee work was significantly lower at h20 than at h40 ($t(8) = 6.60$, $p < 0.016$), and also at h60, ($t(8) = 8.88$, $p < 0.016$). Knee work was also significantly lower at h40 compared to h60 ($t(8) = 7.00$, $p < 0.016$).

The PPMCs among the two ACL injury risk predictor variables and ankle, knee, hip and trunk flexion angles, as well as knee energetics for the vertical height test are reported in Table 2. From Table 2, peak VGRF was significantly and negatively correlated to ankle plantar flexion and knee flexion angle ($r = -77$, $p = 0.02$ and $r = -0.78$, $p = 0.01$, respectively). It is also worth noting from Table 2 that peak PGRF was significantly and negatively correlated to knee flexion angle ($r = -0.72$, $p = 0.03$), as well as significantly and positively correlated to knee power ($r = 0.79$, $p = 0.038$).

Repeated-measure ANOVAs for the horizontal distance test revealed a significant effect of horizontal distance on peak PGRF ($F(2,16) = 28.05$, $p < 0.0001$, partial $\eta^2 = 0.78$, observed power = 1.0), but not on peak VGRF. We also determined a significant effect of horizontal distance on ankle plantar flexion angle ($F(2,16) = 6.70$, $p = 0.008$, partial $\eta^2 = 0.46$, observed power = 0.85), hip flexion angle ($F(2,16) = 6.99$, $p = 0.007$, partial $\eta^2 = 0.47$, observed power = 0.87), and trunk flexion angle ($F(2,16) = 4.15$, $p = 0.04$, partial $\eta^2 = 0.34$, observed power = 0.65). The overall means and standard deviations of the sagittal plane body kinematics, knee power and knee work for the horizontal distance test among all subjects is provided in Table 3. Figure 2 shows the trend in peak PGRF for all subjects as horizontal distance increased. Horizontal distance of landing had a significant and positive effect on peak PGRF, suggesting as horizontal distance of landing increased, peak PGRF increased (Fig. 2). The difference in the marginal means at the three horizontal distances are (0.2, 0.08), which represents the difference in peak PGRF at landing distances 20 to 40 cm and 40 to 60 cm, respectively.

Table 1
The means and standard deviations of each dependent variable among all subjects (Vertical height test).

Dependent variables	Vertical height test		
	Height = 20 cm	Height = 40 cm	Height = 60 cm
Peak VGRF (BW)	3.36 ± 0.71	4.62 ± 1.12	5.35 ± 1.14
Peak PGRF (BW)	0.55 ± 0.19	0.62 ± 0.30	0.79 ± 0.31
Ankle plantar flexion angle (deg)	−0.22 ± 3.52	−2.11 ± 2.44	−2.67 ± 3.54
Knee flexion angle (deg)	−30.88 ± 5.89	−32.92 ± 3.00	−35.89 ± 2.30
Hip flexion angle (deg)	23.31 ± 7.34	19.02 ± 5.53	21.04 ± 6.94
Trunk flexion angle (deg)	13.90 ± 3.18	12.74 ± 2.08	16.38 ± 3.10
Knee power (W/kg)	−5.05 ± 2.95	−10.77 ± 3.03	−15.54 ± 4.40
Knee work (J/kg)	−0.86 ± 0.36	−1.33 ± 0.44	−1.89 ± 0.56

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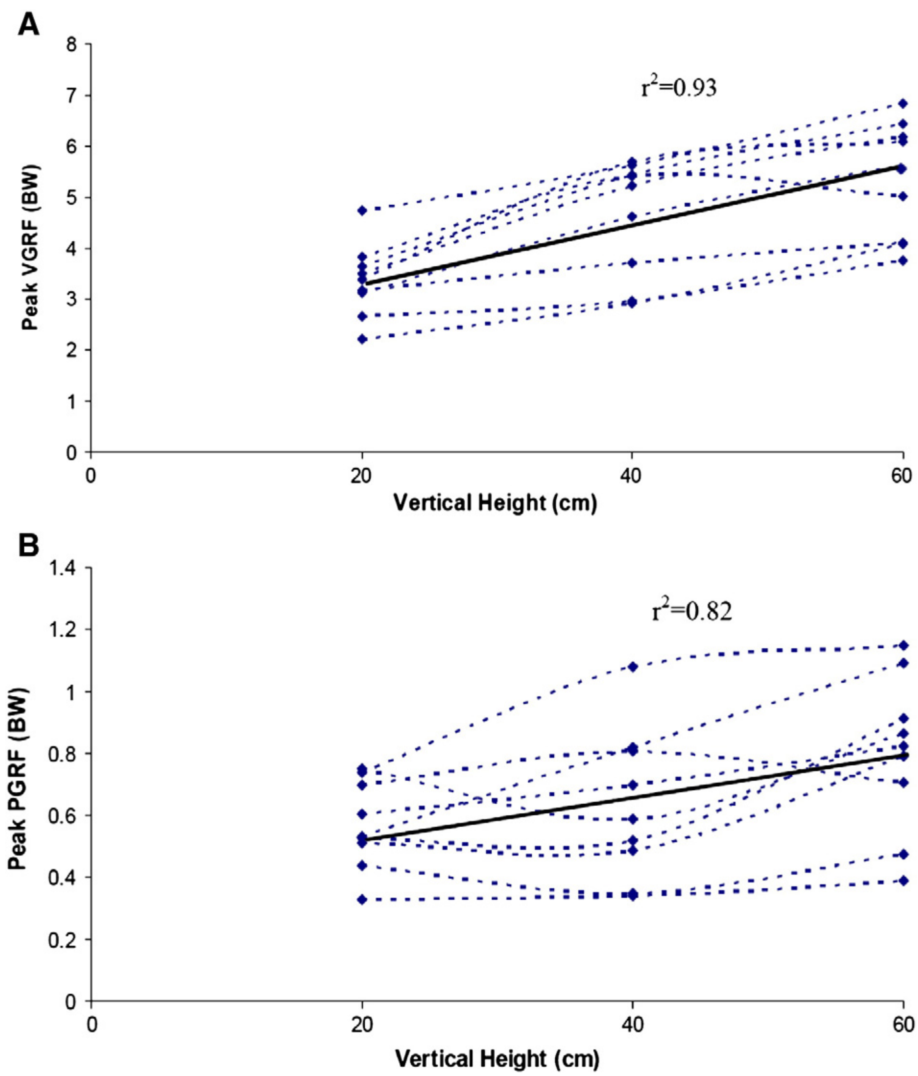

Fig. 1. A: Plot of vertical height and peak VGRF during single-leg landing for all subjects. Solid line is the average slope. B: Plot of vertical height and peak PGRF during single-leg landing for all subjects. Solid line is the average slope.

Pairwise dependent sample t-tests for the horizontal distance test revealed that peak PGRF was significantly different at all three horizontal distances tested. Peak PGRF was significantly lower at d30 compared to d50 ($t(8)=-4.29$, $p=0.003$), and at d70, ($t(8)=-6.69$, $p<0.016$). Peak PGRF was also significantly lower at d50 compared to d70 ($t(8)=-3{,}53$, $p=0.008$). Ankle plantar flexion was significantly higher at d30 compared to d70 ($t(8)=-2.84$, $p=0.015$) and d50, ($t(8)=-2.88$, $p=0.014$). There was no significant difference in ankle plantar flexion at d30 compared to d50. Hip flexion angle was significantly higher at d70 compared to d30 ($t(8)=-3.88$, $p=0.005$). There was no significant difference in hip flexion angle at d30 compared to d50, as well as, d50 compared to d70. Trunk flexion angle was significantly higher at d70 compared to d50 ($t(8)=3.96$, $p=0.004$). There was no significant difference

Table 2
Pearson correlation coefficients of peak VGRF with ankle, knee, hip and trunk flexion angle, as well as knee energetics (Vertical height test).

	Peak VGRF
Peak VGRF (BW)	
Peak PGRF (BW)	r = 0.78 (p =0.01)*
Ankle plantar flexion angle (deg)	r = –0.77 (p =0.02)*
Knee flexion angle (deg)	r = –0.78 (p =0.01)*
Hip flexion angle (deg)	r = –0.14 (p =0.72)
Trunk flexion angle (deg)	r = –0.06 (p =0.88)
Knee power (W/kg)	r = 0.49 (p =0.18)
Knee work (J/kg)	r = 0.04 (p =0.91)

Note: * $p<0.05$.

Table 3
The means and standard deviations of each dependent variable among all subjects (Horizontal distance test).

	Horizontal distance test		
Dependent variables	Distance = 30 cm	Distance = 50 cm	Distance = 70 cm
Peak VGRF (BW)	4.51 ± 1.08	4.68 ± 1.29	4.75 ± 1.45
Peak PGRF (BW)	0.58 ± 0.13	0.78 ± 0.19	0.86 ± 0.20
Ankle plantar flexion angle (deg)	−7.97 ± 2.80	−6.57 ± 1.72	−5.02 ± 1.87
Knee flexion angle (deg)	−40.17 ± 4.89	−37.96 ± 5.87	−38.38 ± 5.55
Hip flexion angle (deg)	22.92 ± 7.60	24.84 ± 8.34	27.31 ± 8.35
Trunk flexion angle (deg)	14.17 ± 3.04	15.01 ± 4.18	17.58 ± 3.80
Knee power (W/kg)	−26.78 ± 6.53	−27.16 ± 6.71	−25.77 ± 7.20
Knee work (J/kg)	−1.42 ± 0.29	−1.52 ± 0.37	−1.55 ± 0.52

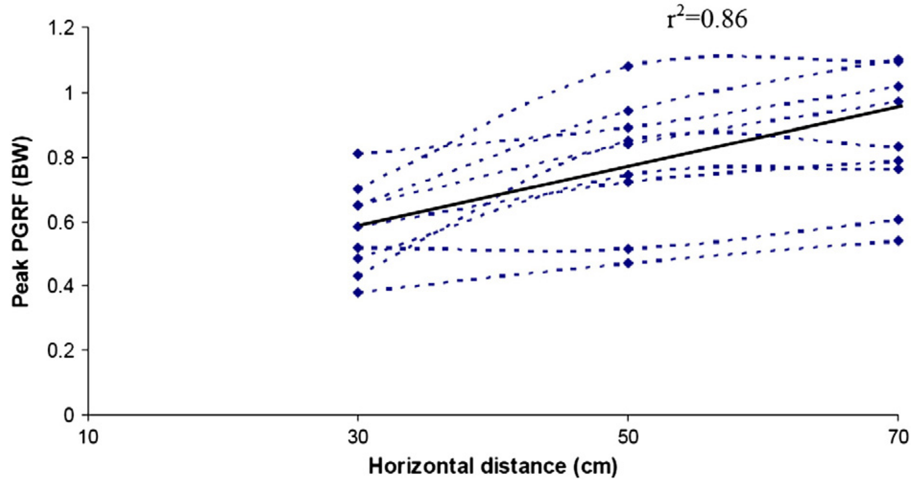

Fig. 2. Plot of horizontal distance and peak PGRF during single-leg landing for all subjects. Solid line is the average slope.

in trunk flexion angle at d30 compared to d50, as well as, d30 compared to d70. Finally, knee work was significantly lower at d50 compared to d70 ($t(8) = -3.53$, $p = 0.008$), with no other significant pairwise comparison.

The PPMCs among the two ACL injury risk predictor variables and ankle, knee, hip and trunk flexion angles, as well as, knee energetics for the horizontal distance test are reported in Table 4. Peak PGRF was significantly correlated with ankle plantar flexion angle ($r = -0.85$, $p = 0.003$), knee power ($r = 0.67$, $p = 0.04$) and knee work ($r = 0.73$, $p = 0.036$). It is worth noting from Table 4 that ankle plantar flexion angle was positively and significantly correlated to knee work ($r = 0.75$, $p = 0.02$). Further, from Table 4, we observed knee flexion angle was positively and significantly correlated to knee power ($r = 0.91$, $p = 0.001$). The time history plots of the ensemble averages of VGRF and PGRF among all subjects at each vertical height and horizontal distance tested are shown in Figs. 3 and 4, respectively. The time history plots of the ensemble averages of ankle, knee, hip and trunk flexion angles among all subjects at each vertical height and horizontal distance tested are shown in Fig. 5A, B, C and D, respectively. In addition, the time history plots of the ensemble averages for knee powers among all subjects are shown in Fig. 6 at each vertical height and horizontal distance tested.

4. Discussion and conclusions

Investigations of how body kinematics function to attenuate impact forces during single-leg landings over increasing vertical heights and horizontal distances may provide new insights into key factors that may mitigate the risk of non-contact ACL injury. Our results showed that an increase in vertical height resulted in a significant increase in peak VGRF and peak PGRF (Table 1). Furthermore, our results showed that an increase in horizontal distance resulted in a significant increase in peak PGRF (Table 3). The finding that an increase in vertical height resulted in increased peak VGRF, which suggests a higher risk of ACL injury, is supported by Yeow et al. [21]. The results of the current study also revealed that single-leg landings did not produce the characteristic bimodal VGRF curve commonly reported for double-leg landings [39,42]. It appears that the demands of single-leg landings from the current study resulted in a rapid increase in VGRFs with a single peak (Fig. 3), which is consistent with the findings by Hargrave et al. [30]. This observation is important as it elucidates the unique nature of single-leg landing studies whose findings cannot be compared with double-leg landing studies.

Table 4
Pearson correlation coefficients of peak PGRF with ankle, knee, hip and trunk flexion angle, as well as knee energetics (Horizontal distance test).

	Peak PGRF
Peak PGRF (BW)	
Peak VGRF (BW)	r = 0.67 (p =0.04)*
Ankle plantar flexion angle (deg)	r = –0.85 (p =0.00)*
Knee flexion angle (deg)	r = –0.29 (p =0.45)
Hip flexion angle (deg)	r = –0.42 (p =0.26)
Trunk flexion angle (deg)	r = 0.54 (p =0.13)
Knee power (W/Kg)	r = 0.67 (p =0.04)*
Knee work (J/kg)	r = 0.73 (p =0.03)*

Note: * $p<0.05$.

A study by Fagenbaum and Darling [31] showed vertical height had a significant effect on knee flexion angles, while a study by Yeow et al. [21] reported no significant effect of vertical height on knee flexion angles. The findings from the current study support the findings of Fagenbaun and Darling [31], that found greater knee flexion angles with increased vertical height (Table 1). Yeow et al. [21] may have found no significant effect of vertical height on knee flexion angle since the subjects performed both single and double-leg landings, where for the single-leg landing tasks the subjects may have landed in a manner that was protective of the ACL. Further, in the current study, subjects had larger body mass (10.92 kg greater) compared to the subjects in Yoew's et al. study [21], suggesting the task from the current study may have been more difficult and resulted in altered landing styles. Results from the current study are also consistent with previous studies that showed a significant effect of vertical height on knee power and knee work [21,24,43]. Vertical height had a significant effect on knee flexion angle, knee power and knee work, which corroborates our first hypothesis.

For the vertical height tests, we found ankle plantar flexion angle was significantly and negatively correlated to peak VGRF (Table 2), and this suggests that increasing ankle plantar flexion may be effective at attenuating peak VGRFs at increased vertical height during single-leg landings. This finding is supported by previous studies [16,24,32,39,44] that found the ankle plantar flexion was effective in absorbing the shock of landing. From Table 2, we also observed that knee flexion angle was significantly and negatively correlated to peak VGRF, which supports our second hypothesis. This finding is consistent with the study by Stacoff et al. [45] who showed that knee flexion angle can be used to reduce the magnitude of the impact loads during landing. In addition, previous studies [22,30,33] have reported that knee flexion plays a key role in force attenuation, subsequently reducing the risk of non-contact ACL injury during single-leg landing, which is consistent with our findings. Table 2 also showed that hip and trunk flexion angles were not associated with peak VGRFs (Table 2). Given there are no single-leg landing studies to

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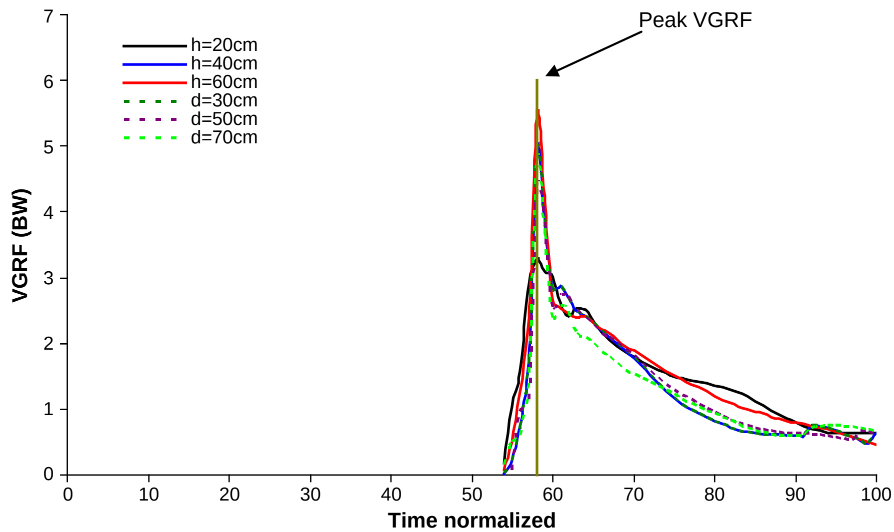

Fig. 3. Time histories of the ensemble average of VGRF at the three vertical heights and horizontal distances tested during single-leg landing among all subjects.

date investigating hip and trunk flexion angles at increasing vertical heights, we could draw no comparison.

For the horizontal distance test, there was a significant effect of horizontal distance on ankle, hip and trunk flexion angle. As well, peak PGRF was significantly and negatively correlated to ankle plantar flexion angles (Table 4). Furthermore, there was no significant correlation between peak PGRF and knee, hip or trunk flexion angle. However, both knee power and knee work were significantly and positively correlated to peak PGRF (Table 4). In addition, for the horizontal distance test, we observed a significant and positive correlation between ankle plantar flexion angle and knee work, as well as, knee flexion angle and knee power (Table 4). These findings suggest that higher ankle plantar flexion and knee flexion angles promote more eccentric work by the knee extensors to dissipate impact energy; a finding supported by a double-leg landing study [11]. Given there are no known single-leg landing studies to date investigating the effect of increasing horizontal distance on risk of non-contact ACL injury, we could make no direct comparisons to confirm the validity of these findings.

Even though this study used a small sample size ($n=9$), we observed sufficiently high statistical power and significance that suggests for some of the results presented, the sample size used was adequate. This is likely because our experimental design used large differences in task demands, we used repeated-measure ANOVAs, and we employed homogenous data (height and weight matched males). While we cannot conclude (given small sample size) that the general male population would exhibit similar sagittal plane body kinematics, knee power, and knee work as reported in the current study, we were able to show that the relationships found have a strong support in terms of partial η^2, r^2 and p values for the subjects tested. Although reasonable correlations were obtained, this study was performed in a controlled laboratory environment where subjects could plan for these tasks and as such may not be representative of maneuvers experienced during sports. Even though subjects performed at least two trials at each landing height and distance, reliability data could not be determined given only one trial was post processed and there was not enough data points for the nine subjects tested. As well, even though single-leg landings performed in the

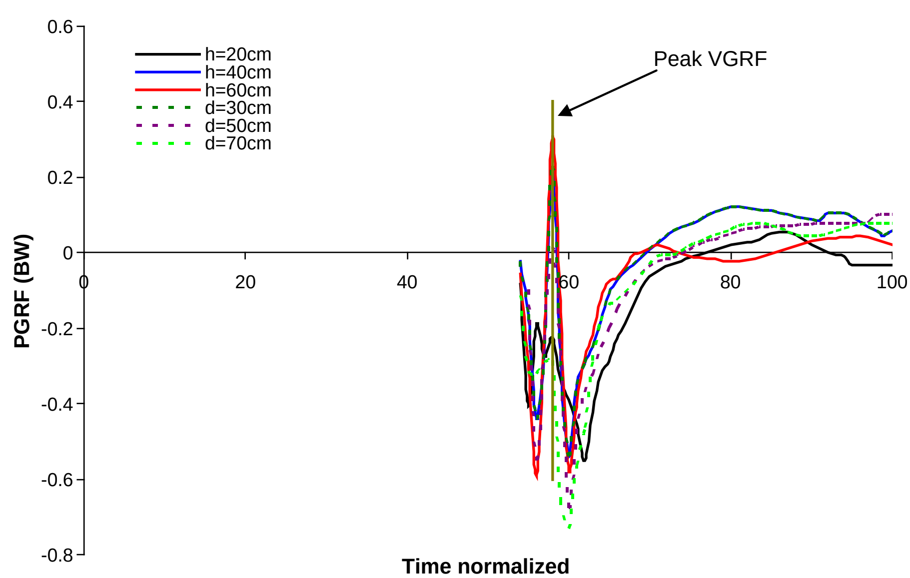

Fig. 4. Time histories of the ensemble average of PGRF at the three vertical heights and horizontal distances tested during single-leg landing among all subjects.

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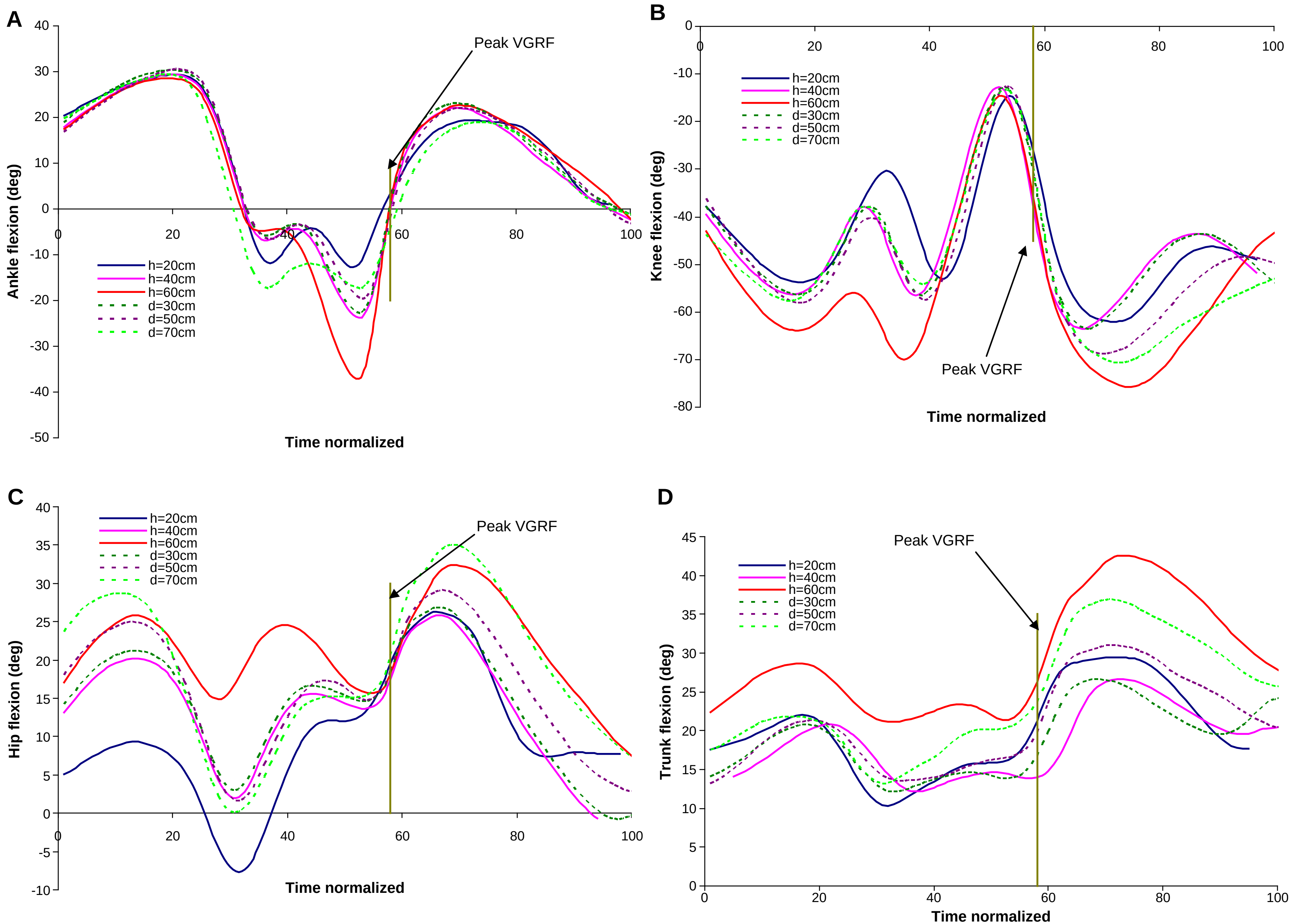

Fig. 5. A: Time histories of the ensemble average of ankle flexion angles at the three vertical heights and horizontal distances tested during single-leg landing among all subjects. B: Time histories of the ensemble average of knee flexion angles at the three vertical heights and horizontal distances tested during single-leg landing among all subjects. C: Time histories of ensemble average of hip flexion angles at the three vertical heights and horizontal distances tested during single-leg landing among all subjects. D: Time histories of the ensemble average of trunk flexion angles at the three vertical heights and horizontal distances tested during single-leg landing among all subjects.

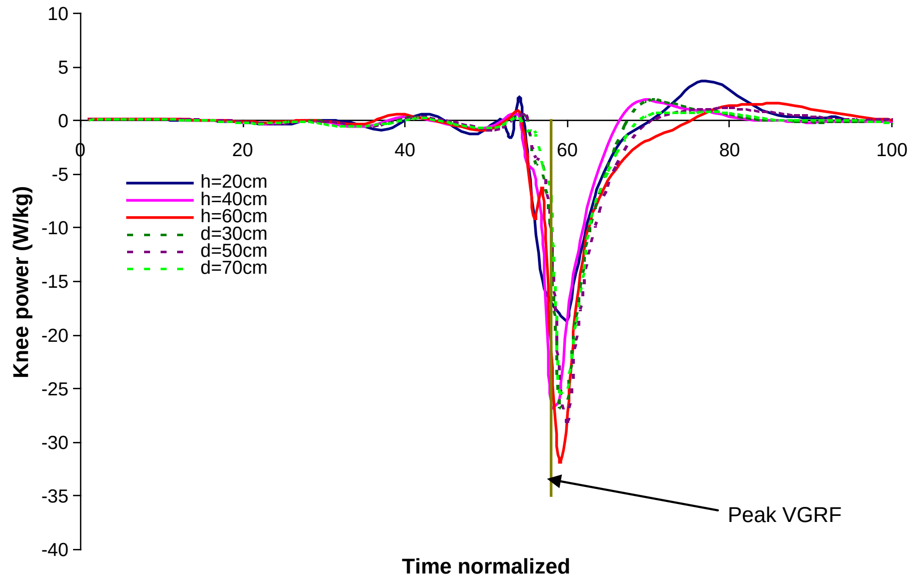

Fig. 6. Time histories of the ensemble average of knee power at the three vertical heights and horizontal distances tested during single-leg landing among all subjects.

current study were sagittal plane dominant, out-of-plane movements commonly involved in sports and not captured in this study may be important contributors to the risk of non-contact ACL injury. In the present study, the single-leg landing tasks investigated entailed little out-of-plane motion and therefore it is likely that valgus loading implicated to increase the risk of non-contact ACL injury may not be influential. However, an area of future research remains single-leg landings from varying vertical height and horizontal distance that involves out-of-plane motion such as landing to the medial or lateral aspect of the knee. Furthermore, since we did not measure ACL loads during the single-leg landing tasks, the results and discussions stemming from this study are only based on what is known about the relationship between ACL loads and GRFs. Further studies addressing the aforementioned limitations, as well as accounting for the effect of gender during single-leg landings from increasing vertical heights and horizontal distances are needed to shed more light on the biomechanics of non-contact ACL injury during single-leg landing.

This study investigated the relationships among vertical heights and horizontal distances on peak GRFs, sagittal plane body kinematics, and knee energetics, as well as related these findings to risk of non-contact ACL injury during single-leg landing. Within the findings and limitations of this study, we observed that vertical height had a significant effect on peak VGRF, peak PGRF, knee flexion angle, trunk flexion angle, knee power and knee work. We also observed that horizontal distance had a significant effect on peak PGRF, as well as ankle, hip, and trunk flexion angles. Results from PPMCs showed that larger ankle plantar flexion and knee flexion angles as well as knee energetics may aid in attenuating peak VGRF and peak PGRF, which may promote more appropriate muscle firing patterns to protect the ACL.

Conflict of interest statement

All authors declare that there are no direct or indirect financial interests involved with the content of this paper.

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